REPORT DOCUMENTATION PAGE	*Form Approved* *OMB No. 0704-0188*

1. REPORT DATE *(DD-MM-YYYY)*	2. REPORT TYPE	3. DATES COVERED *(From - To)*
1994	Conference Proceeding	9/29/1992 - 2/2/1993

4. TITLE AND SUBTITLE

Loads Induced in the Lumbar Spine of Seated Restrained Humans by Sideward (+Gy) Impact

5a. CONTRACT NUMBER

5b. GRANT NUMBER

5c. PROGRAM ELEMENT NUMBER

6. AUTHOR(S)

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5d. PROJECT NUMBER

62202F-7231-2401

5e. TASK NUMBER

5f. WORK UNIT NUMBER

7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)

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8. PERFORMING ORGANIZATION REPORT NUMBER

AFRL-HE-WP-SR-1998-0004

9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)

US Air Force Research Laboratory
Human Effectiveness Directorate
2800 Q Street, Bldg 824
Wright-Patterson Air Force Base, OH 45433-7947

10. SPONSOR/MONITOR'S ACRONYM(S)

11. SPONSOR/MONITOR'S REPORT NUMBER(S)

12. DISTRIBUTION/AVAILABILITY STATEMENT

Approved for public release; distribution is unlimited.

13. SUPPLEMENTARY NOTES

AsMA Conference, San Antonio, TX, 1994

14. ABSTRACT

20060921194

15. SUBJECT TERMS

16. SECURITY CLASSIFICATION OF:			17. LIMITATION OF ABSTRACT	18. NUMBER OF PAGES	19a. NAME OF RESPONSIBLE PERSON
a. REPORT	b. ABSTRACT	c. THIS PAGE			John R. Buhrman
					19b. TELEPHONE NUMBER *(Include area code)*
Unclassified	Unclassified	Unclassified	UU	11	937-255-3121

HSMA Conference, San Antonio TX, 1994 Strzelecki, J.P.

LOADS INDUCED IN THE LUMBAR SPINE OF SEATED RESTRAINED HUMANS BY SIDEWARD (+Gy) IMPACT.

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ABSTRACT: INTRODUCTION: The parameters used to calculate the Dynamic Response (DR - the currently accepted injury criteria for multi-axis impact) for sideward impact are based on a very limited data set. In addition, the current model for spinal injury due to z axis acceleration is based on displacement and cannot account for the effects of off-axis loads. METHODS: Seated, restrained human volunteers were subjected to sideward impacts ranging from 0.031 to 0.250 seconds duration and amplitudes of from 4 to 7 G. Loads were measured at all restraint points and used to calculate dynamic coefficients for a model of upper body response and lumbar spine shear loading. RESULTS: Lumbar spine shear loads can be predicted using a second-order lumped parameter model with a natural frequency of 58 rad/sec and damping ratio of 0.45. CONCLUSIONS: Combining these results with similar models for the x and z axes and correlating them with injury data will allow a comprehensive model of lumbar spinal injury to be produced.

KEYWORDS: Dynamic Response, Acceleration, Injury

Jeni: this will be a footnote at the bottom of the first page of the following text: The author is a mechanical engineer at the Crew Escape and Protection Branch, Armstrong Laboratory, WPAFB

The currently accepted criteria for impact injury in the USAF is known as the Dynamic Response (DR). The DR is three mutually independent perpendicular single degree of freedom lumped parameter second order systems whose displacements due to impact accelerations are correlated to injury potential. The dynamic coefficients used to define the model for the Y axis are based on the results of a single experiment with a very limited scope (2).

This study had four main goals: First, to refine or validate the model parameters for the Y axis by testing over a broad range of acceleration durations and amplitudes. Second, to investigate the linearity of the human response. Third, to calculate the loads induced in the lumbar spine by sideward impacts, and finally to model the inertial response of the upper body for use in design of flight controls for the fourth generation ejection seat. The last two items rely on the same method of analysis, therefore the calculation of the lumbar loads will be presented in detail and results given for the upper body inertial response.

A specially designed seat was mounted on a sled with seated human volunteers facing a direction perpendicular to the applied acceleration. The volunteers wore a PCU-15/P parachute harness and HGU-55/P flight helmet. The acceleration pulse was provided by a pneumatic ram and the sled was constrained to coast down track rails. The seat provided metal restraints at the feet, knees, and hips which effectively limited subject motion to the torso

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#11916

and head. The subject was further restrained by a lap belt and shoulder straps anchored to the seat. All contact and anchor points were instrumented with load cells. To prevent the volunteers' arms from flailing during impact, very loose velcro loops were placed around each thigh. The volunteers were instructed to hold the loops with palms up and not to brace their arms against their thighs. Figure 1 displays the test seat.

>Figure 1 here<

The tests were conducted in three phases. The first phase consisted of a moderate impact to serve as an orientation to the program. In the second phase (Cells A through D), the volunteers were subjected to impacts of increasing amplitude but of constant duration. In the third phase (Cells E through I), the amplitude was held constant but the pulse duration was varied. Table 1 displays the test matrix.

TABLE 1: TEST MATRIX FOR THE SIDEWARD IMPACT STUDY

Seat Acceleration	Nominal Pulse Duration (milliseconds) 31	64	82	150	210	250
4 G				X,A		
5 G				B		
6 G	E	F	G	C	H	I
7 G				D		

In the past, a velcro strap wrapped around the volunteer's chest served as an attachment for an accelerometer array. Model dynamic coefficients were then calculated based on the measured accelerations. Unfortunately, this is not a rigid mounting technique and results in a large amount of scatter in the data.

To avoid these problems, a different approach was taken. First, the time histories of all reaction forces acting on the volunteer were continuously summed, giving the total body inertial response due to the impact as a function of time.

The lower body is effectively immobilized; its contribution to the total body inertial response is the lower body weight multiplied by the sled acceleration. The lower body (legs and hips below the lumbar spine) is taken to be 50% of total body weight. The basis for this assumption is a military study of anthropometry (3) which used stereophotographic techniques to obtain mass distribution data on body segments.

Subtracting the lower body inertial response (delayed 6 msec to account for plasticity at the lower body restraint points) from the total body inertial response gives the dynamic response of the upper body. The six msec delay was determined empirically by overlaying plots of the seat acceleration and total inertial response time profiles. It represents an average value for cells E through I. Using a free body diagram, one can then calculate the loading on the lumbar spine as a function of time. Figure 2 illustrates the method.

>Figure 2 here<

For the twelve volunteers who completed Cells A, B, C, and D, the variation of peak lumbar load with peak sled acceleration proved highly linear. For each subject, the peak calculated lumbar loads at these cells were taken as data points (with the origin providing a fifth data point) and a least squares straight line calculated. The correlation coefficients (r2) had a mean of 0.9850 and standard deviation of 0.0113.

<jeni, the r2 is r with superscript 2>

To determine the coefficients of a second-order lumped parameter model of lumbar loading as a function of seat acceleration, the following method was employed:

First, for each of cells E, F, G, C, H, and I, the peak lumbar load was plotted against volunteer upper body weight and a least-squares straight line was determined for the data set. Figure 3 is a sample.

>Figure 3 here<

From the equation of this line the expected lumbar load for a 220 lb volunteer was calculated. The natural frequency and damping ratio required for a model which would match this lumbar load for each pulse duration were then determined. The best fit was provided by a model with a natural frequency of 58 rad/sec and damping ratio of 0.45. The study on which the DR is based gave the natural frequency as 58 rad/sec and damping ratio of 0.09. The low damping was attributed to poor coupling between the subject and the seat pan; the subjects in that study were unable to brace to the degree possible in this study. Thus what is likely seen in the earlier study is a rigid body oscillation. Figure 4 shows the analytical model. Not shown in the model is a 25 msec delay on the acceleration required for a good fit. Table 2 shows the quality of the fit.

>Figure 4 here<

TABLE 2: Model fit for 220 lb volunteer subjected to a 6 G sideward impact.

Cell	Pulse Duration (msec)	Lumbar Load from Test Data (lb) (w/95%tile conf.)	Lumbar Load from Dynamic Model (lb)
E	31	516+/-35	498.
F	64	788+/-50	788.
G	82	811+/-64	810.
C	150	681+/-68	743.
H	210	683+/-61	727.
I	250	678+/-88	694.

The lumbar load calculated from the test data for Cell C was judged to be an anomaly caused by the scatter pattern in the data. Also, the cells were run in alphabetical order as required by the Human Use Committee; the Cell C results may reflect a lack of experience in bracing on the volunteers' part.

Several possible solutions would provide a good fit to the data presented in table 2; as a cross-check on the parameters selected, overlays of the calculated and predicted load time histories were examined to insure the parameters chosen were correct. Figure 5 is a sample, and shows that the model fit is good in both pulse width and location of the peak. It was these overlays that determined the required time delay in the model.

>Figure 5 here<

The Advanced Dynamic Anthropomorphic Manikin (ADAM) is the current state-of-the-art in human analogs for impact (1). The ADAM has a lumbar load cell which can be used to measure forces directly. Three tests were run at each test cell, and the results compared with the model predictions for a volunteer of the same weight (218 lb). The results are given in Table 3. The values presented for ADAM are the average of the three tests.

TABLE 3: Comparison of lumbar loads measured in ADAM and predicted by the model

Cell	Pulse Duration (msec)	ADAM (lb)	Dynamic Model (lb)
E	31	78.	493.
F	64	169.	781.
G	82	193.	803.
C	150	236.	736.
H	210	268.	720.

The table shows that ADAM's lumbar load cell cannot be relied upon to give accurate results. The reason is that in a sideward impact the plastic "waist" of the manikin provides an alternate load path as the body segments come in contact.

A model was formulated for the upper body inertial response which is intended for use in flight control programming for the fourth generation ejection seat.

The response was calculated in the same manner as for the lumbar load calculation (fig. 2). In this case the model has a natural frequency of 30 rad/sec, damping ratio of 0.40, gain of 1.70, and delay of six msec on the input acceleration. Even with very light damping a gain factor would be required to achieve the measured force amplitude; therefore the damping ratio was chosen to correspond closely with the lumbar load model damping ratio and also provide a good fit on time-domain plots of inertial response. The gain was then chosen to make up the difference between model and experimental results. The gain probably accounts for rotational effects and the higher acceleration of the head and perhaps the arms.

Table 4 is a comparison of inertial responses extrapolated from the experimental data and calculated from the model.

TABLE 4: Model fit for 220 lb volunteer subjected to a 6 G sideward impact.

Cell	Pulse Duration (msec)	Inertial Load from Test Data (lb) (w/ 95% conf)	Inertial Load from Dynamic Model (lb)
E	31	615+/-55	457.
F	64	1061+/-72	1022.
G	82	1208+/-74	1197.
C	150	1416+/-91	1416.
H	210	1392+/-108	1391.
I	250	1374+/-132	1328.

The model has the same form as Figure 4, with the following changes: the seat acceleration is multiplied by 1.70 * 0.5 * wt/386., the spring and damping coefficients are different, the time delay is 6 msec versus 25 msec in the previous model, and the output is upper body inertial load.

Figure 6 is a sample of the fit obtained with the model for one test.

<Figure 6 here>

CONCLUSIONS: Lumbar spinal loads and upper body inertial response can be predicted quite well with simple second order lumped parameter models with constant coefficients. The response varies quite linearly with applied seat pan acceleration. Finally, the ADAM manikin lumbar load cell does not provide a good indication of lumbar loads to be expected in humans in sideward impacts.

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2. Brinkley JW. Acceleration Exposure Limits for Escape System Advanced Development. SAFE Journal, 1985, Vol 15, No 2.

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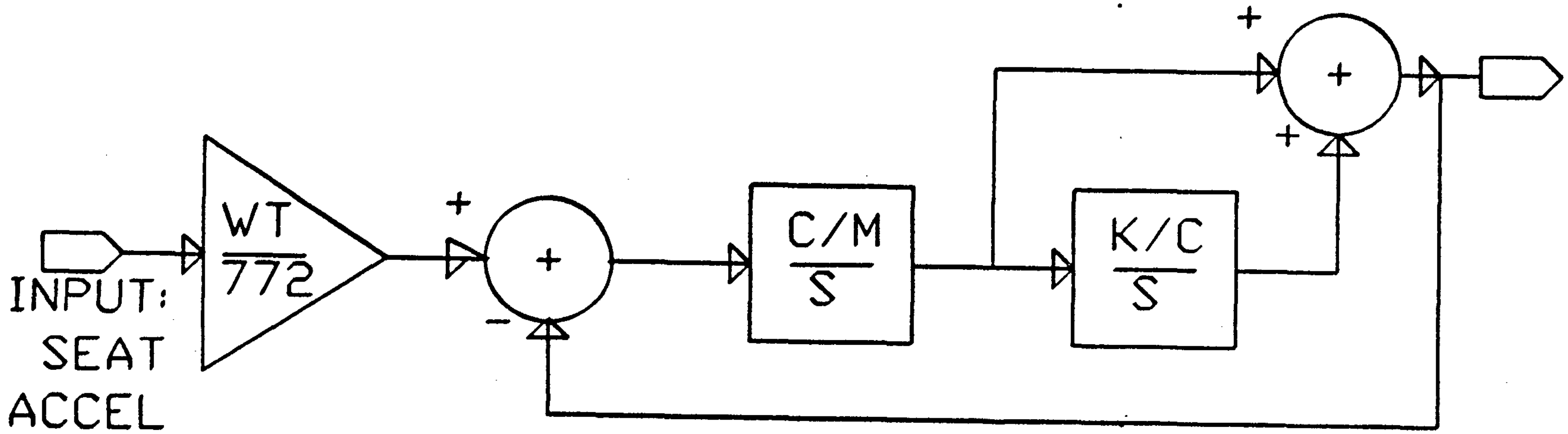
SECOND ORDER LUMPED PARAMETER MODEL

FOR $\ddot{Y}$=SEAT PAN Y ACCEL:

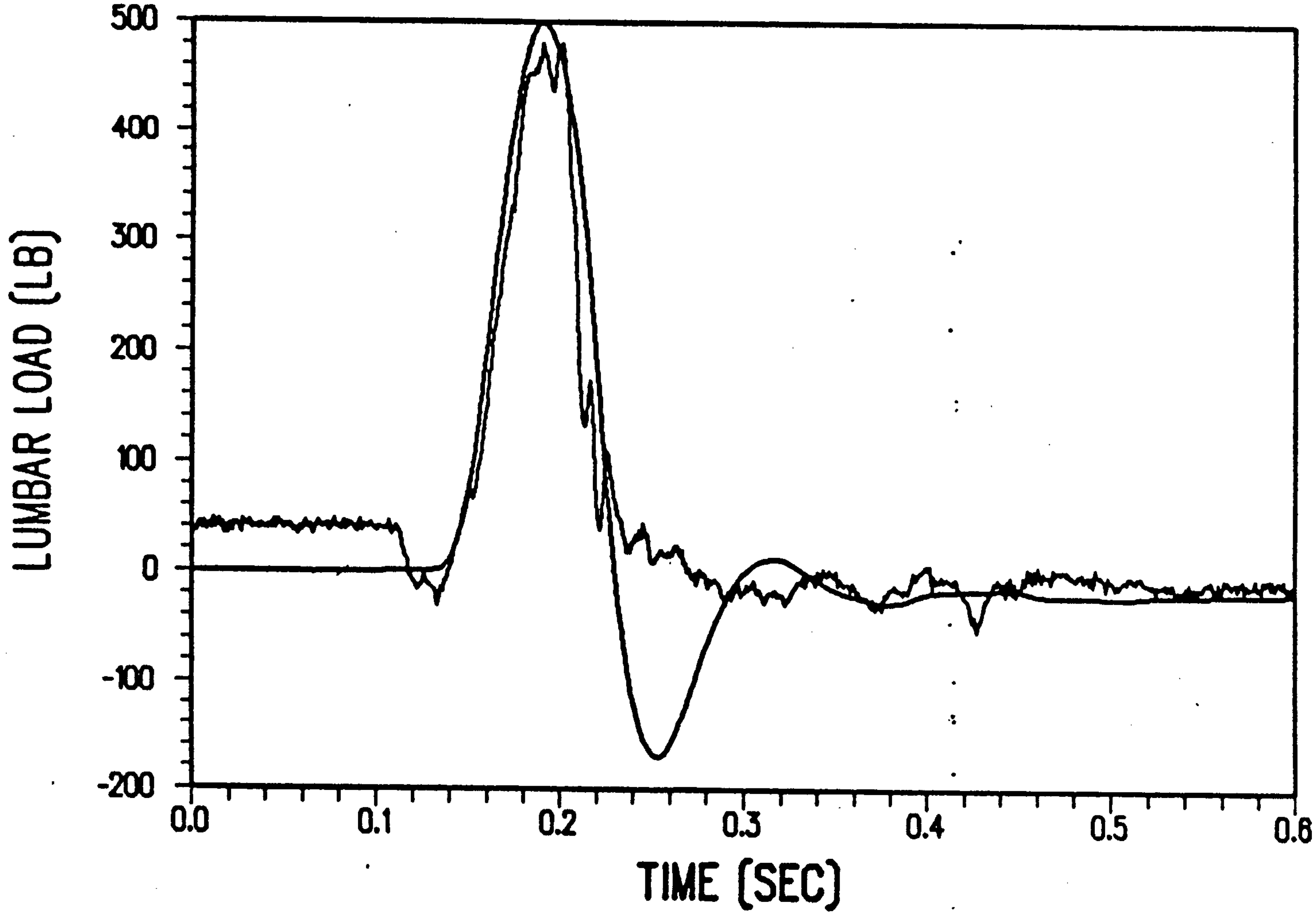
$M\ddot{X}+C\dot{X}+KX=C\dot{Y}+KY$

LET $Z=X-Y$, THEN

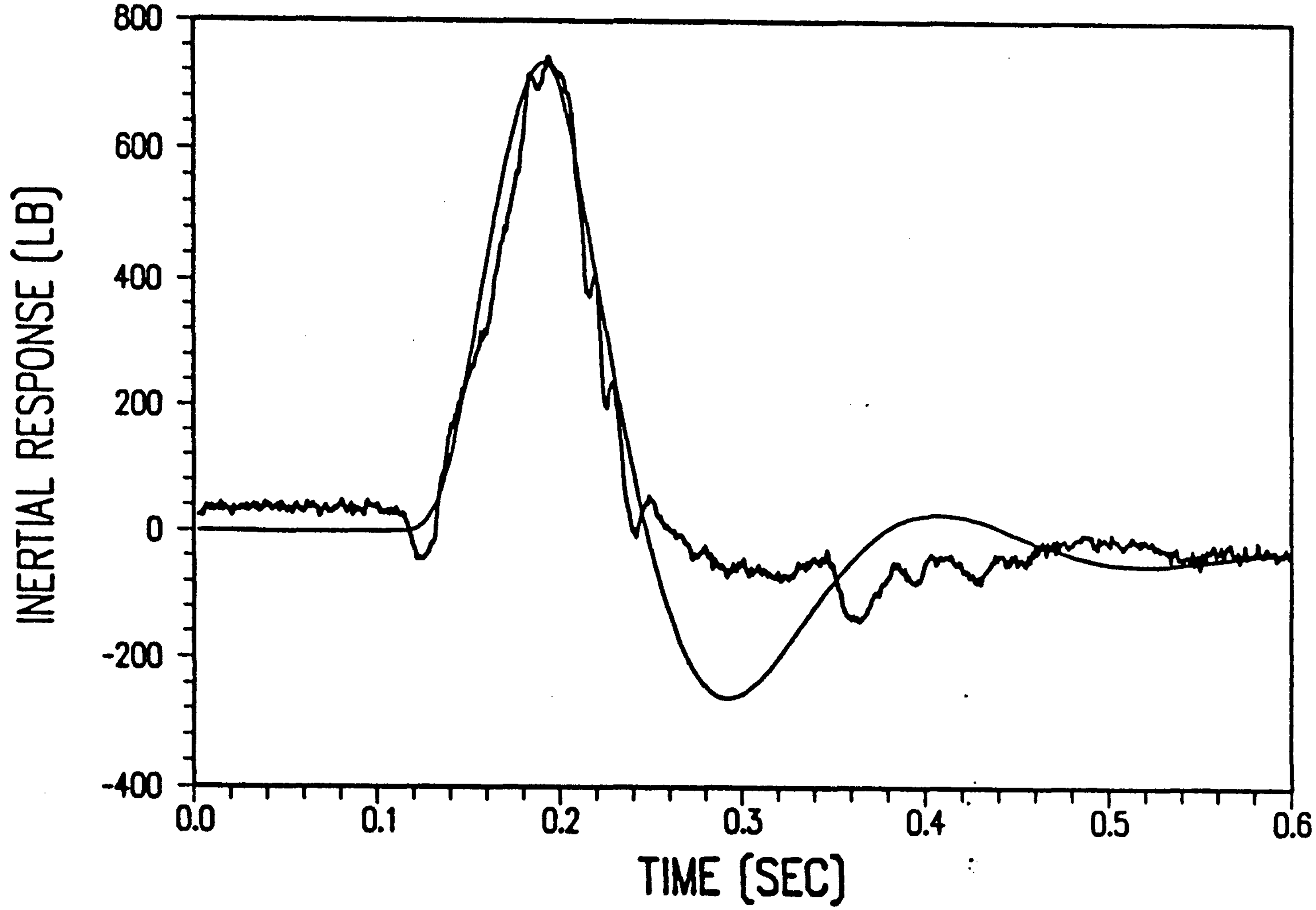
$-M\ddot{Z}=(C\dot{Z}+KZ)+M\ddot{Y}$

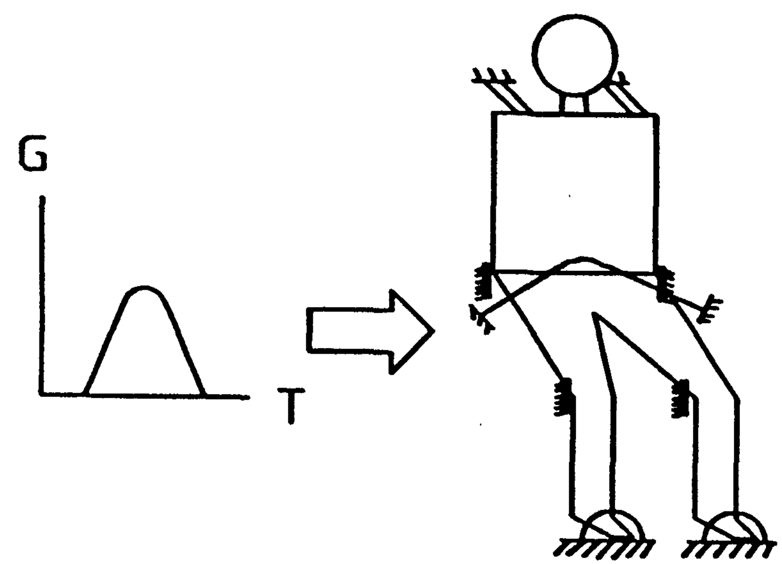


HGY CELL G TEST 4247
MODEL VS DATA



HGY UPPER BODY INERTIAL RESPONSE
CELL G (82 msec)

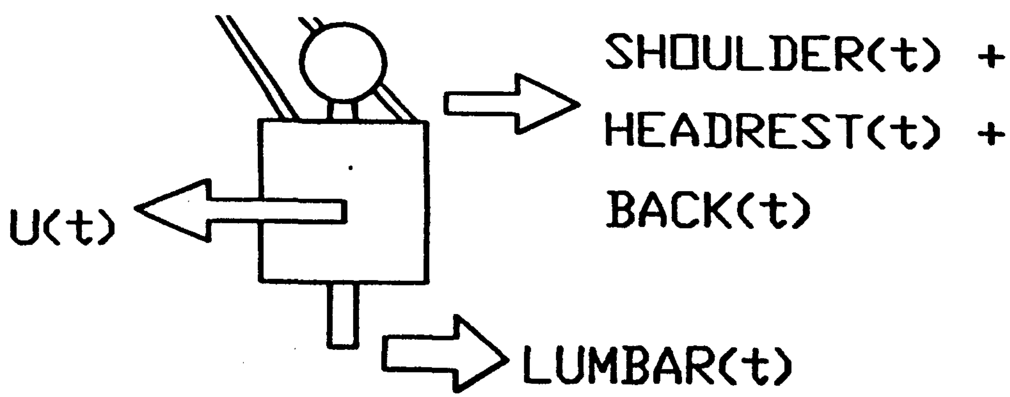




Y-AXIS ONLY

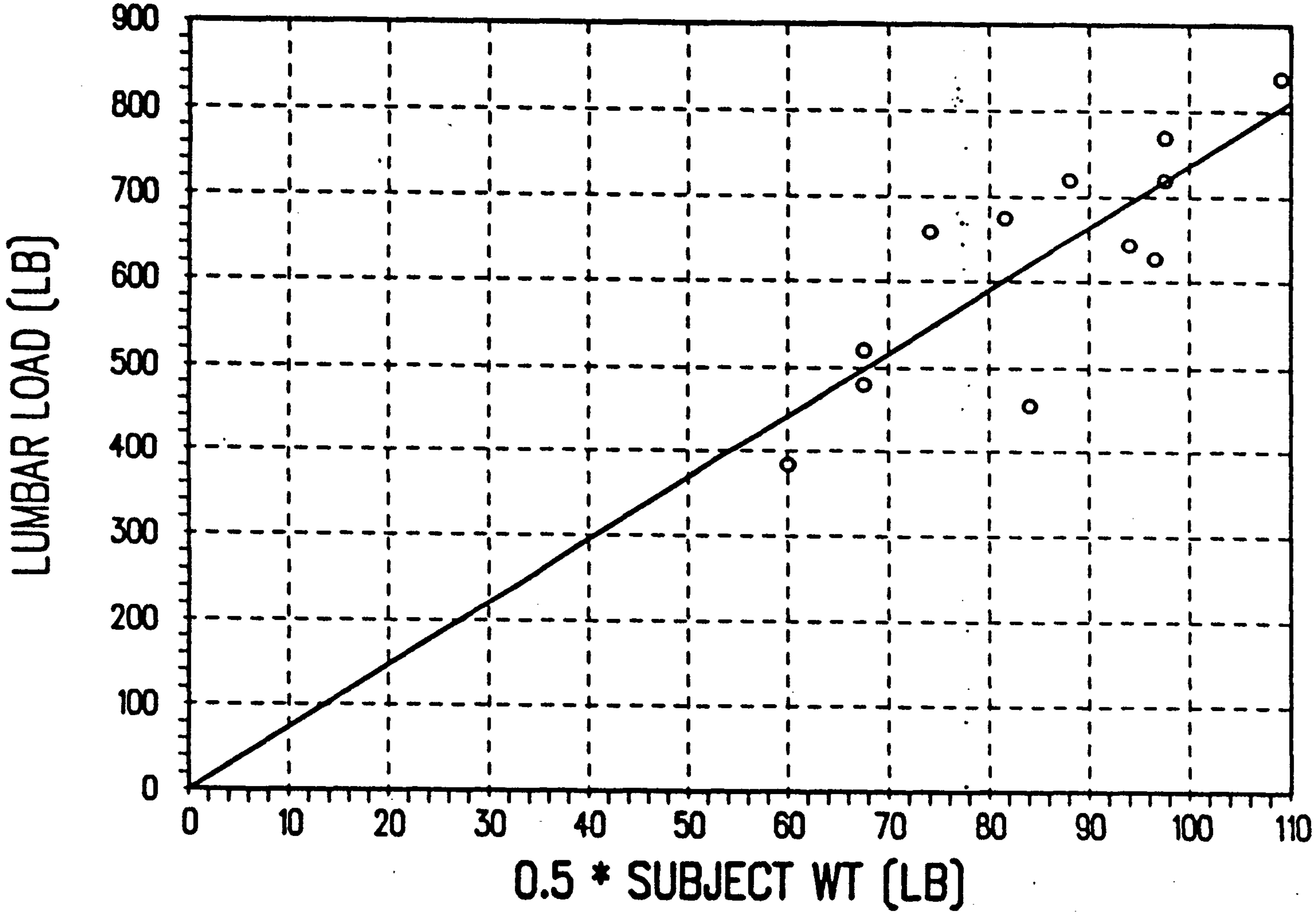
TOTAL INERTIAL LOAD$_Y$ I(t)=
HEADREST(t) + SHOULDER(t)
+ BACK(t) + PAN(t) + HIP(t)
+ KNEE(t) + FOOT(t)

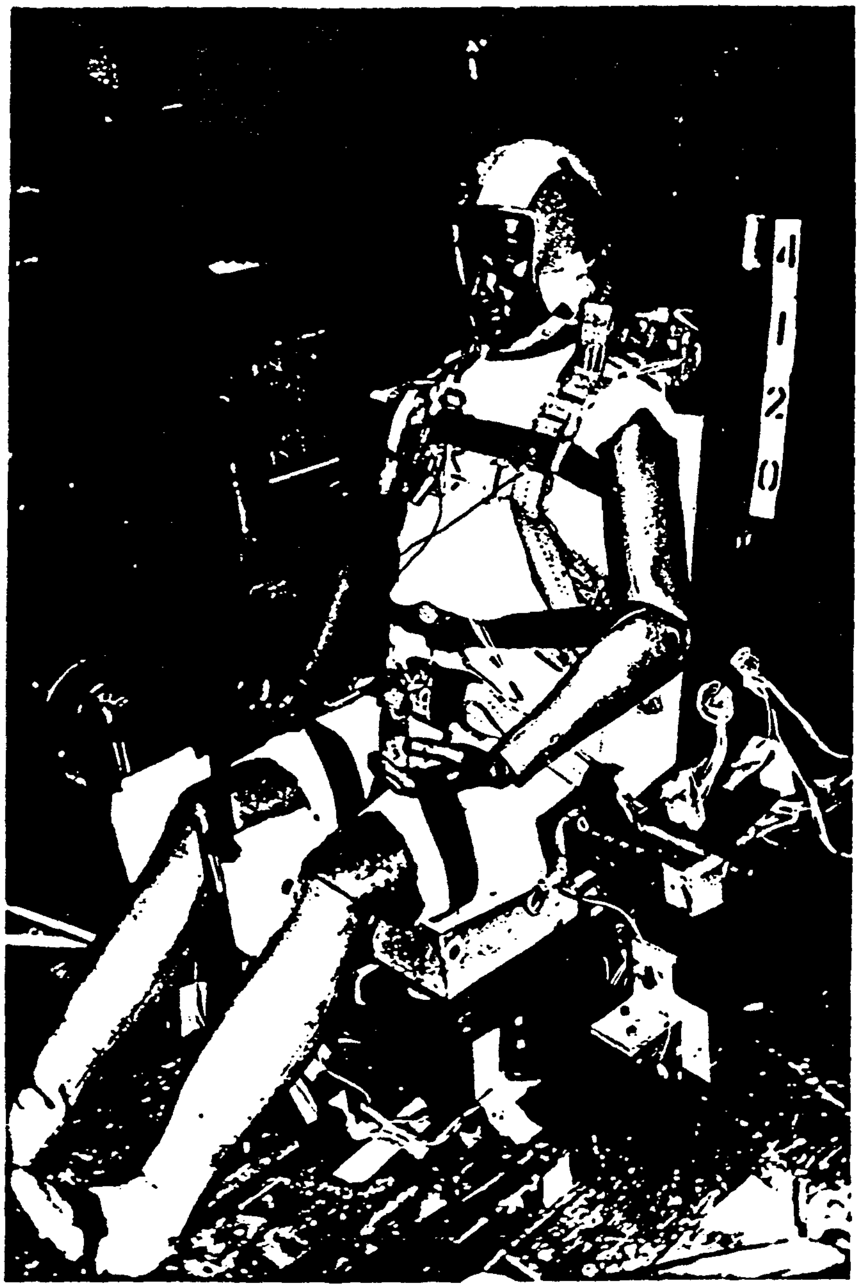
UPPER BODY LOAD U(t) =
I(t) - (0.5 * WT * G(t))



FREE
BODY
DIAGRAM

HGY CELL G
Y = 7.3703 * X

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